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Automatic Generation Of A Computational Model For Monopolar Stimulation Of Cochlear Implants.

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Purpose: Cochlear implants have the potential to significantly improve severe sensorineural hearing loss[1]. However, the outcome of this technique is highly variable and depends on patient-specific factors. We previously proposed a method for patient-specific electrical simulation after CI, which can assist in surgical planning of the CI and determination of the electrical stimulation pattern [2]. However, the virtual implant placement and mesh generation were carried out manually and the process was not easily applied automatically for further cochlear anatomies. Moreover, in order to optimize the implant designs, it is important to develop a way to stimulate the results of the implantation in a population of virtual patients.
In this work we propose an automatic framework for patient-specific electrical simulation in CI surgery. To the best of our knowledge, this is the first method proposed for patient-specific generation of hearing models which combines high-resolution imaging techniques, clinical CT data and virtual electrode insertion. Furthermore, we show that it is possible to use the computational models of virtual patients to simulate the results of the electrical activation of the implant in the cochlea and surrounding bone. This is an important step because it allows us to advance towards a complete surgical planning and implant optimization procedure.

Methods: The proposed framework is shown in Figure 1. Detailed cochlear surface models were built from high-resolution images of cadaveric samples of the inner ear, using a Scanco μCT 100 system (Scanco Medical AG, Switzerland) and advanced image processing tools [3]. A statistical shape model (SSM) of the cochlea was constructed [2-3]. The SSM is able to generate new virtual cochlear shapes by means of sampling methods, and a fitting procedure is used to obtain a patient-specific surface from clinical CT images of the patient [3].
In order to simulate the electrode insertion, the centreline of the cochlea is computed, which corresponds to the center of the scala tympani. The electrode array geometry was provided by the manufacturer Med-EL based on the Flex28 design with 12 contacts. The virtual electrode insertion algorithm reorients the electrode mesh along the cochlear centreline using the parallel transport frame approach [4]. This is a fast and robust algorithm which allows controlling most of the most relevant insertion parameters, such as the percentage of insertion.
Once the surface mesh of the whole hearing model is created, including the cochlea, electrode and surrounding bone, we proceed with the volumetric mesh generation. The meshing framework was developed in Matlab and using open-source toolboxes [5]. All independent meshes were merged automatically, ensuring the lack of intersection. In order to ensure the convergence in the subsequent finite element electrical simulations, the mesh quality was assessed. The quality was quantified by its aspect ratio expressing the element quality within a range from 0 to 1, meaning nearly degenerated elements and perfect tetrahedron element, respectively.
Nerve fibres were added to the model, to be able to couple electrical stimulation and neural response [2]. An automatic method was developed for the generation of the nerve fibres along the patient-specific cochlea. It allowed defining their number in a parametric fashion, and using as initial points those contained on the cochlear line located on the basilar membrane.
For the cochlear implant simulation, we modeled a monopolar stimulation protocol, where each electrode is activated and stimulates with an intensity of 1mA, and the surrounding bone acts as a reference. The protocol was implemented in a

template simulation using the quasi static electrical regime of the Maxwell equations as implemented in the Elmer multiphysics solver. The boundary conditions were defined only once in the template of the SSM and then propagated to all the instances of virtual patients created by sampling the SSM. This enables our approach to potentially scale to thousands of patients because no manual intervention is required. All simulations ran until convergence in the cluster at our Institution.

Results: The meshing framework (Fig.1) was tested on 7 cochlear anatomies instantiated from the SSM. The meshing process took around 2 minutes for each model generation. Mesh quality was very similar for all cases, around 0.807±0.127 for each mesh. In Table 1 the degree of convergence of the 7 computational models created, showing that all were equally stable. In Figure 1 we show the voltage and volume current field calculated after simulation. This summarizes the electrical activity undergoing in the cochlea as an effect of the implant stimulation. The seven simulated patients present differences both in the size of the scalae and in the spacing between the turns. Those differences are caused by the sampling process of the SSM that effectively creates new patient geometries. However, the value of the maximum potential field stays similar between the various patients, even if the anatomy changes. This is expected, because it is mainly the shape of the inserted electrodes that determines the potential field and this was not changed during our experiments. We expect to have deep changes in the activation patterns of the auditory nerves for each patient instead, but more work is needed to verify this hypothesis.

Conclusions: We presented an automatic patient-specific mesh generation and defined the monopolar clinical stimulation protocol for cochlear implants on a template model and then propagated it to seven computational models of virtual patients. Specifically, this patient-specific hearing model allows to provide a detailed evaluation of the outcomes of CI, thus moving a step closer to in-silico testing of the outcomes of CI surgery.

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Tables

Patient	p1	p2	p3	p4	p5	p6	p7
Convergence	0.38e-12	0.62e-12	0.68e-12	0.14e-10	0.15e-11	0.19e-11	0.72e-12

Table 1: Convergence values for the seven virtual patients

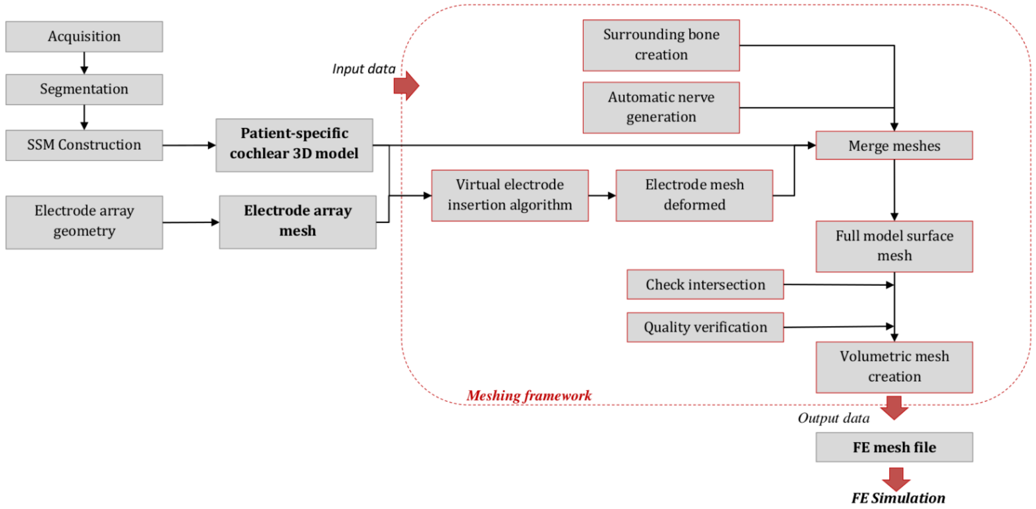

Figure1. General pipeline for the patient-specific model generation. Marked in red the automatic meshing framework proposed in this abstract.

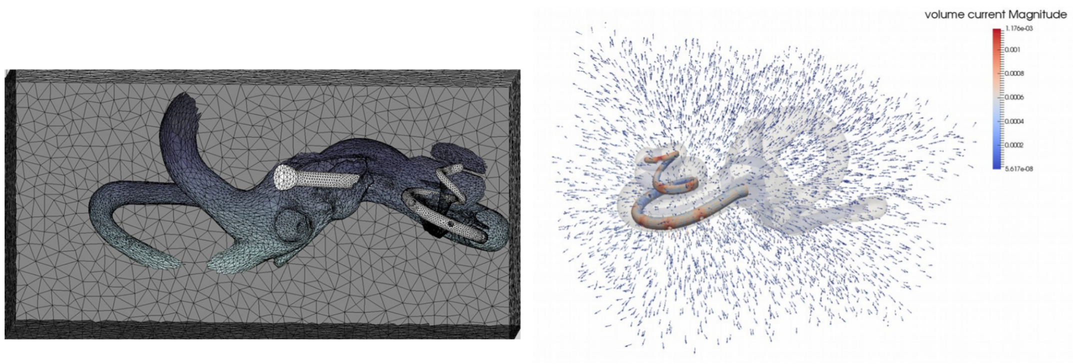

Figure 2. Results of the mesh generation pipeline. Left: Final mesh obtained. It includes the cochlea anatomy, the nerve fibres, the surrounding bone and the electrode array inserted. Right: Voltage and volumne current field. This shows the electrical activity undergoing in the cochlea as an effect of the implant stimulation